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Evaluation of Control Interfaces for Active Trunk Support

Stergios Verros¹, Nauzef Mahmood, Laura Peeters, Joan Lobo-Prat, Arjen Bergsma, Edsko Hekman, Gijsbertus J. Verkerke, and Bart Koopman

Abstract—A feasibility study was performed to evaluate the control interfaces for a novel trunk support assistive device (Trunk Drive), namely, joystick, force on sternum, force on feet, and electromyography (EMG) to be used by adult men with Duchenne muscular dystrophy. The objective of this paper was to evaluate the performance of the different control interfaces during a discrete position tracking task. We built a one degree of freedom flexion–extension active trunk support device that was tested on 10 healthy men. An experiment, based on the Fitts law, was conducted, whereby subjects were asked to steer a cursor representing the angle of the Trunk Drive into a target that was shown on a graphical user interface, using the above-mentioned control interfaces. The users could operate the Trunk Drive via each of the control interfaces. In general, the joystick and force on sternum were the fastest in movement time (more than 40%) without any significant difference between them, but there was a significant difference between force on sternum on the one hand, and EMG and force on feet on the other. All control interfaces proved to be feasible solutions for controlling an active trunk support, each of which had specific advantages.

Index Terms—Active trunk support, control interfaces, biomechatronics.

I. INTRODUCTION

THE MOST common form of muscular dystrophy in humans is Duchenne muscular dystrophy (DMD), affecting 1 in every 3,500 boys [1]. DMD causes progressive degeneration of muscles, leading to progressive loss of muscle strength [2]. The mean life span used to be about 20 years of age but, due to improved health-care practices and ventilation,

this has now increased to about 25–30 years [3], [4]. However, several studies have shown that people with DMD have lower Health Related Quality of Life (HRQoL) compared to healthy controls [5], [6]. As the dystrophy progresses over the years, impairments in body functions increase progressively. Thus, people with DMD become dependent on caregivers, especially with a reduction in arm function.

Several dynamic arm supports have been developed with the aim to increase the independence of people with decreased arm functions [7]. It has been shown that arm reach can be greatly increased when it is augmented with trunk motion [8]. Currently, there are no active trunk support devices and as DMD-patients have little voluntary trunk control in more advanced disease stages there is a demand for active trunk support. As a first step, we designed the Trunk Drive. The Trunk Drive is basically an experimental active trunk orthosis that supports and stabilizes flexion and extension of the trunk in a seated position.

Intention detection control interfaces are needed to operate active assistive orthotic devices, to provide the communication between the users intended motion and the device. Several control interfaces have been used to this end by others [9]–[13]. Force-based and electromyography (EMG) control interfaces have been proposed as the most promising strategies to control an arm support for people with DMD [10]. Hand joysticks are used as control interfaces by people with disabilities in order to control powered wheelchairs and external robotic arms [11], [12]. Therefore, we decided to include joystick, force and EMG based modalities in this study.

Two force-based control interfaces were tested. The first one measured the interaction force between the sternum and the Trunk Drive and the other one measured the interaction forces underneath the feet. The force between the sternum and the Trunk Drive can be changed by flexing or extending the trunk. The force underneath the feet can be changed by exerting isometric flexing or extending moments around the knees. To control a single degree-of-freedom (DoF), EMG signals from two muscles (usually an antagonistic muscle pair) need to be recorded [10]. We measured antagonistic pairs of leg muscles in subjects to obtain EMG control signals since it is shown that control tasks can be performed better using leg muscles than trunk muscles [14].

As a first step in the design of a trunk support prototype, which can be used in combination with an arm support to increase arm reach, we compared the performances of

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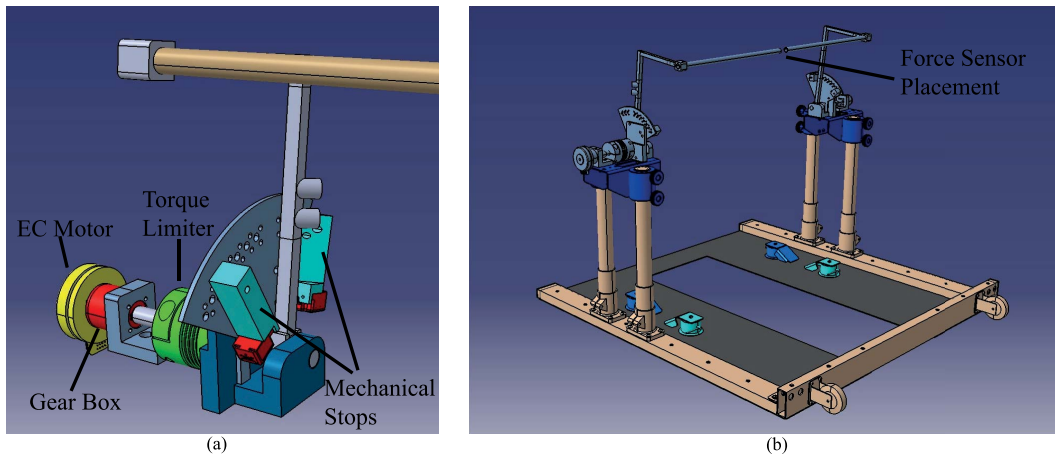


Fig. 1. Components of the Trunk Drive system. (a) Two EC motors with torque limiters are placed on the left and right at the L3 level of the spine to assist flexion and extension. Two mechanical stops ensure that the range of motion is between the allowable limits. (b) The rectangular structure is the floor plate and the subject sits between the vertical bars. The subject's trunk is attached to the horizontal bar with a plastic cup that encircles the trunk (not shown in the picture).



Fig. 2. Adult man using the Trunk Drive system at zero degree flexion angle. The trunk cup works as an adjustable connection mechanism between the Trunk Drive and the subject. The subject can operate an emergency button to stop the experiment in an unpleasant situation.

the four interfaces on the Trunk Drive. We were motivated from previous studies where it was shown that people with DMD with weakened muscles can control an arm exoskeleton using different control interfaces [10]. In this Fitts' law based experiment, we used 10 healthy volunteers as test subjects.

II. METHODS

A. Design and Actuation

An experimental test set-up was designed (Trunk Drive) to investigate the four control interfaces when performing a simple movement, namely one DoF flexion and extension in a seated position (Fig. 1, 2). The Trunk Drive consists of an aluminum alloy frame that encloses the trunk of the user from two sides on the frontal plane. The front horizontal bar interfaces at sternum level with the trunk of the user through a plastic pad. The two side bars are attached to two identical motor (Maxon EC 90, gearbox reduction 91:1) shafts that drive the flexion-extension movement through bearing and torque isolators to prevent frictional loss and misalignment of the shaft axes. The actuated rotational point of the Trunk Drive was kept aligned with the L3 level of the spine of the user by shifting the height of the Trunk Drive. The minimal

flexion angle was set at 0° (trunk oriented vertically) and the maximum angle was set at 40° flexion. To prevent movements exceeding this range of motion, limit switches were used in conjunction with the mechanical stops.

B. Sensors

An optical encoder (Maxon Encoder MILE 512-64000 CPT, 2 channels) was used to measure the angular positions of the motors. The user's intention was detected by using either the signal from a joystick, a force sensor at the sternum, a force sensor under the feet or a set of EMG electrodes. We used a one DoF joystick whose spring stiffness could be adjusted. The force-based control interface on the sternum measured the interaction forces between the human and the device with a six DoF load cell (ATI mini45) located at the height of the sternum between the trunk cup and the metallic bar. Only the forces acting perpendicular to the sternum were taken into account. The force-based control interface under the feet measured the interaction forces between the feet and the ground. The feet were fixed to the force plate and only the horizontal forces in the forward and reverse direction, resulting from knee flexion and extension moments, were taken into account. Trunk flexion was coupled with the backward forces of the feet and vice versa. The reason behind this decision was to reflect standing up from a seated position. The flexion activation signals were measured from the lower tibialis anterior and the extension activation signals from the gastrocnemius. Two differential surface electrodes (Trigno, Delsys, USA) per muscle were placed parallel to the muscle fibers according to SENIAM recommendations [15].

C. Signal Acquisition and Control Hardware

The sensor signals were sent to a computer (xPC Target, MathWorks Inc., USA) by means of a data acquisition card (PCI-6229; National Instrument Corp., USA) which made analog-to-digital conversions with a sampling frequency of 1 KHz and a 16-bit resolution. The controller was also

connected to the computer and sent the calculated voltage to the motor driver (ESCON 90/50) through the same data acquisition card which then provided the appropriate current to the motor (Maxon EC 90 brushless).

D. Signal Processing

The real-time signal processing was adapted from Lobo-Prat *et al.* [10]. The joystick was a simple potentiometer with a resistance of 1k and 5V-feed from the data acquisition card. The following equation describes the joystick's signal processing::

$$J_{vol}(i) = J_{sen}(i) - J_{res} \quad (1)$$

$$u_{joy}(i) = \begin{cases} \frac{J_{vol}(i)}{J_{mvi,f}}, & \text{if } J_{vol}(i) < 0 \\ \frac{J_{vol}(i)}{J_{mvi,e}}, & \text{if } J_{vol}(i) > 0. \end{cases} \quad (2)$$

J_{sen} is the sensed intention of movement, J_{res} is the average signal amplitude of a resting period of two seconds, J_{vol} is the voluntary movement and u_{joy} is the joystick control signal. $J_{mvi,f}$ and $J_{mvi,e}$ are the maximum voluntary inclination (MVI) of the joystick when pushing the joystick forward (flexion of trunk) and pushing the joystick backwards (extension of trunk) for two seconds, respectively. Finally, the (i) represents the i^{th} time step of the signal.

Regarding the force-based control interface on the sternum, it is necessary to distinguish the voluntary forces of the user from external forces such as gravity or joint stiffness and to compensate for these. The external force is obtained before the actual measurement by measuring the sternum interface forces during a slow flexion of the trunk (0.05 rad/s) while the subject is fully relax. The compensated force $F_{com}(\theta)$ is a function of the trunk angle (θ) and it is subtracted from the sensed forced $F_{sen}(i)$ to realize the intended movement as is shown in the following equations:

$$F_{vol,\theta}(i, \theta) = F_{sen}(i) - F_{com}(\theta) - F_{res} \quad (3)$$

$$u_{force}(i, \theta) = \begin{cases} \frac{F_{vol}(i, \theta)}{F_{mvc,f}}, & \text{if } F_{vol}(i) < 0 \\ \frac{F_{vol}(i, \theta)}{F_{mvc,e}}, & \text{if } F_{vol}(i) > 0. \end{cases} \quad (4)$$

Where $u_{force}(i, \theta)$ is the force control signal, $F_{mvc,f}$ and $F_{mvc,e}$ are the two seconds abdominal (flexion) and ilio-costalis (extension) maximum voluntary contraction (MVC), F_{res} is the average signal amplitude of a resting period of 2 seconds and J_{vol} is the voluntary movement. The force plate signal processing can be described as:

$$FP_{vol}(i) = FP_{sen}(i) - FP_{res} \quad (5)$$

$$u_{forceplate}(i) = \begin{cases} \frac{FP_{vol}(i)}{FP_{mvi,f}}, & FP_{vol}(i) < 0 \\ \frac{FP_{vol}(i)}{FP_{mvi,e}}, & FP_{vol}(i) > 0. \end{cases} \quad (6)$$

Envelope detection was applied to the raw EMG signal with a high-pass Butterworth filter at 40 Hz, a full wave rectifier and a low pass Butterworth filter at 2 Hz [16]. Furthermore, additional signal processing was performed to normalize the

TABLE I
ADMITTANCE CONTROL PARAMETERS

Control Interface	Virtual Mass	Virtual Damping
joystick	0.3	1
force on sternum	0.1	0.5
force on feet	0.3	1.5
EMG	0.5	4

two signals from the agonist and antagonist muscles. In the following equation, $E_{nor,k}(i)$ represents the normalized EMG and $u_{emg}(i)$ the control signal:

$$E_{nor,k}(i) = \frac{E_{env,k}(i) - E_{res,k}}{E_{mvi,k}} \quad (7)$$

$$u_{emg}(i) = E_{nor,t}(i) - E_{nor,g}(i) \quad (8)$$

Subscript k represents the abbreviation of the tibialis (t) and gastrocnemius (g) muscles. $E_{env,k}$ is the envelope of the raw EMG, $E_{res,k}$ is the average of the signal amplitude in a rest period of two seconds, $E_{mvi,k}$ is the maximum value during 2 seconds of maximum voluntary isometric contraction.

E. Control

The control architecture consists of two levels: higher and lower level control. The higher level control is a second order admittance model with virtual mass A and virtual damping B where A, B were tuned with the different control interfaces in order to achieve low movement time and low overshoot. The values were kept constant between the subjects and they were chosen in such a way that the time to complete a task and the overshoot were minimized in a pre-experimental procedure with 2 subjects that were not included in the experiment. The admittance model generates the reference position from the intention detection signal. The position reference is followed by the lower level control. The lower level controls the position of the two motors using one PD controller for each motor. The PD values of the controller were tuned manually (Fig. 3, Table I).

III. EXPERIMENTAL DESIGN

The experiment was based on the approach described by Fitts [17], who identified a predictive model describing human speed accuracy trade off in a tapping task. A one-dimension serial position-tracking task was presented to the subject by means of Python custom-made Graphical User Interface (GUI) on a 1050 Å– 1680 pixels display.

The user has to steer a cursor (yellow circle in Fig. 4) from a home position (blue circle) to a target (red circle). The cursor was coupled with the encoders of the left motor. The Index of Difficulty (ID, 9) from Shannon's form [18] was used to characterize targets. A target was presented at one of the three different locations at a distance of 395, 791 and 1583 pixels from the cursor starting position. The target radius remained constant at 70 pixels plus 30 pixels for cursor correction [19]. The GUI is shown in Fig. 4.

$$ID = \log_2 \left(\frac{D}{R} + 1 \right) \quad (9)$$

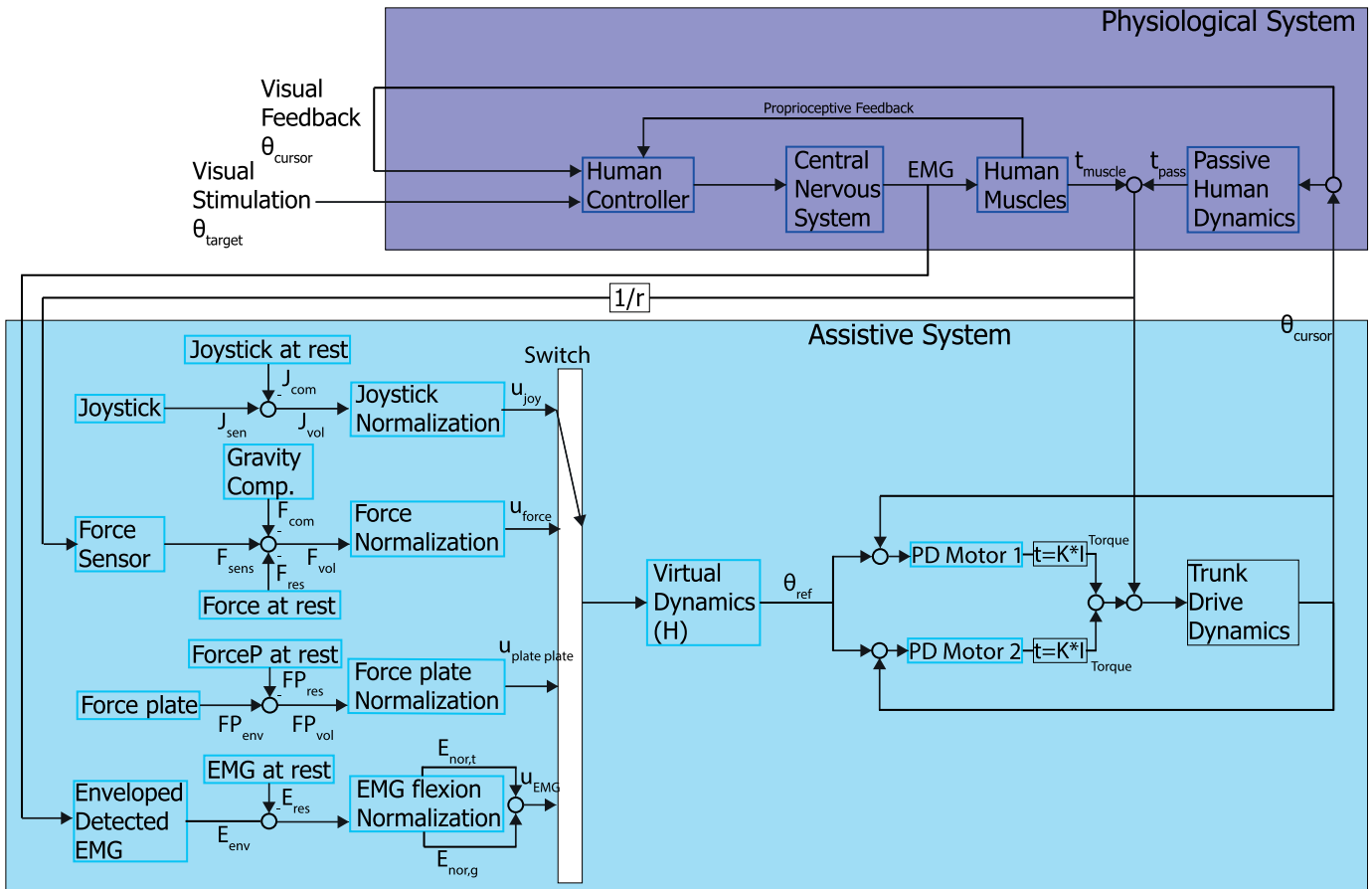


Fig. 3. Control Diagram of the Trunk Drive system with the physiological system. After the visual stimulation θ_{target} , the subjects had to steer the θ_{cursor} into the target by using one of the four control interfaces. The $u_{joy}/force/force\ plate/EMG$ created a reference angle θ_{ref} to be followed by the low level PD control.

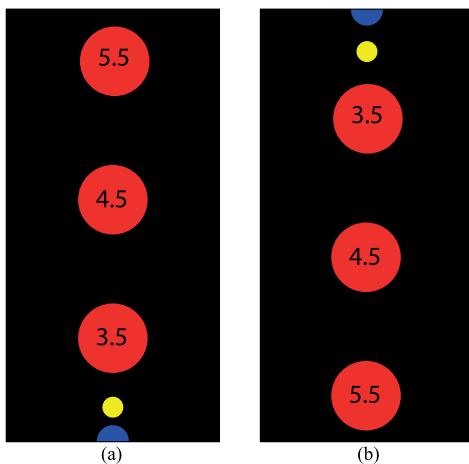


Fig. 4. Representation of the Graphical User Interface. A yellow cursor starts in the blue circle, ID increases with larger distances from the home position. The pixel/angle relation is 42.

A. Task

After familiarization themselves with the device using the joystick with their dominant hand, subjects were asked to perform flexion/extension movements by using the four different control interfaces. At initiation, the trunk was at zero degree

flexion angle and the cursor was at the home position 1 at the bottom of the screen. To indicate the start of a trial, the word 'Go' was displayed on the screen together with a target, and the subject had to move the cursor into the target and keep it there for two seconds. If the target movement was successful after a dwell time of two second, a blue target was shown at the top of the screen indicating home position 2 (permitted maximum flexion position). When the word 'Go' appeared, the subject had to move the cursor again into a new target and remain there for a dwell time of two seconds to complete the extension trial. After the extension task was completed, the blue home position 1 appeared and a new flexion task was started.

B. Performance Metrics

Performance metrics were used to assess the control interfaces by giving a more detailed picture of their advantages and disadvantages. The following performance metrics were used (Fig. 5):

Movement time (MT): the time needed to complete the task after the reaction time and without a dwell time.

Throughput (TP): also known as information transfer rate in bits/s, measures how much information can be conveyed from a subject through a particular command source; it was

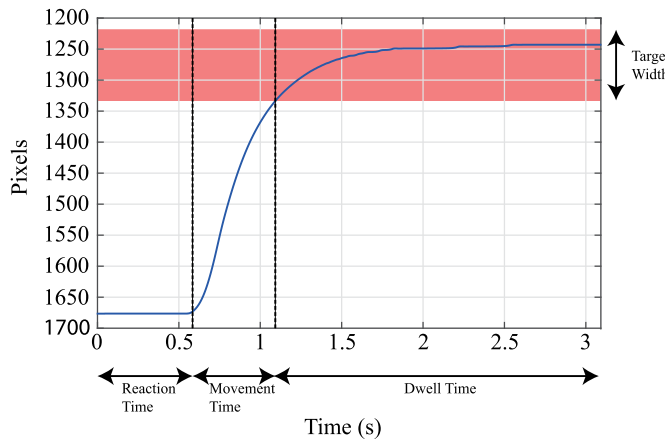


Fig. 5. Typical step like response of one trial with the joystick as control interface.

computed using (10).

$$TP = \frac{ID}{MT} \quad (10)$$

Reaction Time (RT): is the time counted between the appearance of the target and attaining 2% of the maximum speed of the trial.

Path Efficiency (PE): measures the straightness of the cursor and it is computed as a percentage of the straight line.

$$PE = 100\% \frac{Straight_Distance}{Actual_Distance} \quad (11)$$

Overshoot (OS): indicates how many times the cursor left the target before the end of the dwell time of two seconds, whereby only the first time that the cursor left the target is counted and divided by the number of targets.

A questionnaire was administered at the end of the experiment to evaluate the users' experiences with the different control interfaces. The subjects had to evaluate the control interfaces by answering which control interface was: most accurate, fastest, easiest to control, most exhausting, easiest to install and, most comfortable. Finally, an overall preference had to be given.

C. Protocol

After detailed explanation of the tasks, the subjects were placed into the Trunk Drive. The joystick was the first control interface to be tested (Fig. 6). The other three control interfaces were randomized for every subject. The subjects were given five minutes before an experiment was started to familiarize themselves with a new control interface. The subjects had to complete eight blocks consisting of nine trials of flexion and nine trials of extension (three for every target) in randomized order for each control interface. The first three blocks were for practicing and only the last five were analyzed. During the experiment, the subjects were instructed to move to the target as fast as possible without overshooting.

D. Participants

In total, 10 healthy male subjects (27.6 ± 2.45 years) gave their informed consent and participated in this study. The medical Ethics Committee of Radboud University Medical Center approved the study and the design protocol (NL53143.091.15).

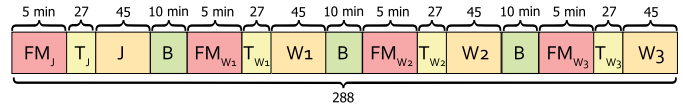


Fig. 6. Trial sequence of the experiment. FM stands for familiarization, T for training, J for joystick and B for break. W1, W2, W3 represent the random force sequences on the sternum and the feet and the EMG. 288×2 flexion and extension trials were performed.

TABLE II

REGRESSION COEFFICIENTS OF THE ID VERSUS MOVEMENT TIME AND THE r VALUE. I, S AND r STANDS FOR INTERCEPT, SLOPE AND r VALUE RESPECTIVELY

	$I_{flexion}$	$S_{flexion}$	$r_{flexion}$	$I_{extension}$	$S_{extension}$	$r_{extension}$
J	-0.5685	0.4206	0.833	-0.2905	0.3574	0.753
F_s	-0.3045	0.3158	0.829	-0.5368	0.3617	0.827
F_f	-1.6909	0.8272	0.814	-0.8387	0.5577	0.819
E	-2.0397	0.9620	0.831	-3.1751	1.2394	0.804

E. Statistics

Statistical analysis was performed between the control interfaces and, in the case of MT, each ID was also statistically analyzed. The average of each subject per control interface was calculated for every metric. Since not all the data were normally distributed, a Friedman test was performed. Post-hoc Wilcoxon signed rank with Bonferroni correction was followed by 6 pair-wise comparisons. A significant level of $\alpha = 0.0083$ and $\alpha = 0.0017$ was indicated by * and ** respectively. The r values represent the square root of the coefficient of determination that was calculated on the linear regression.

IV. RESULTS

The following abbreviation will be used: J for joystick, F_s for force on sternum, F_f for force on feet and E for EMG. The MT results were analyzed per ID since there were differences between the control interfaces for each ID.

A. Movement Time

1) **Linear Relationship:** The mean MT was calculated per person per flexion and extension and it was plotted versus ID. The r values between ID and MT were above 0.814 for flexion and above 0.753 for extension (see Table II and Fig. 7).

2) **Comparison MT Per ID:** Flexion: The Friedman test showed a significant difference in movement time between control interfaces in each ID ($p < 0.001$, $nDOF = 3$, $\chi^2_{ID=3.5} = 22.68$, $\chi^2_{ID=4.5} = 26.04$, $\chi^2_{ID=5.5} = 24.12$). When ID = 3.5 (Fig. 8 and Table III), the joystick was significantly faster than EMG and force on feet. Force on sternum was also significantly faster than force on feet and EMG. When ID = 4.5, force on sternum was significantly faster than joystick, force on feet and EMG. The Joystick was also significantly faster than force on feet and EMG. Finally, when ID = 5.5, force on sternum was significantly faster than force on feet and EMG. The Joystick was also significantly faster than force on feet and EMG.

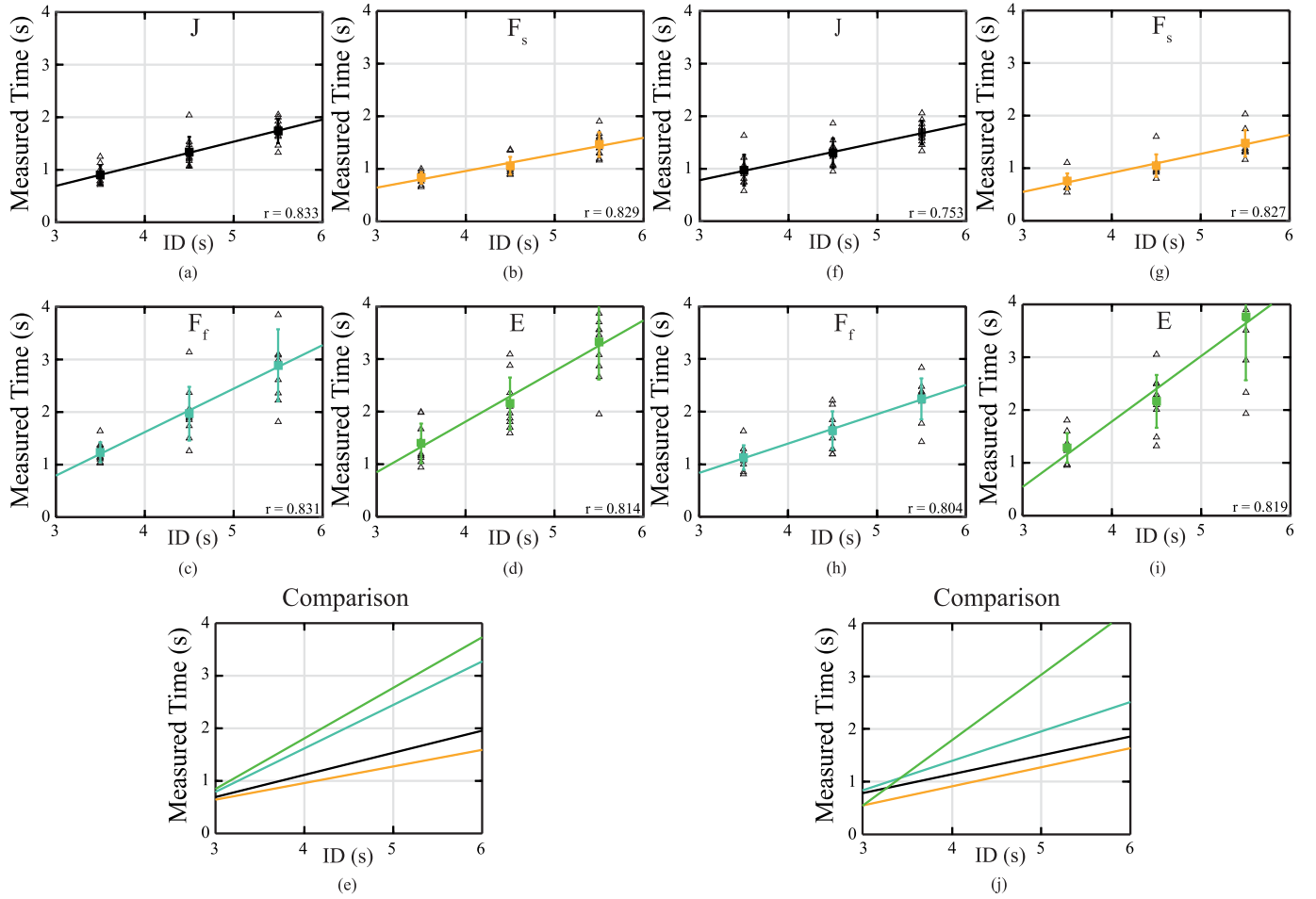


Fig. 7. Linear regression plots of the MT versus ID. (a) J flexion (b) F_s flexion (c) F_f flexion (d) E flexion (e) comparison flexion (f) J extension (g) F_s extension (h) F_f extension (i) E extension (j) comparison extension. Each triangle represents the average time per ID per subject. The error bars represent the mean per ID for 10 subjects with 1 standard deviation. The legend applies to all Figures.

TABLE III
MT AND STANDARD DEVIATION

	MT _{flexion}	SD _{flexion}	MT _{extension}	SD _{extension}
$J_{(3.5)}$	0.90	0.19	0.97	0.29
$J_{(4.5)}$	1.34	0.29	1.30	0.28
$J_{(5.5)}$	1.74	0.23	1.69	0.23
$F_{s(3.5)}$	0.83	0.11	0.75	0.15
$F_{s(4.5)}$	1.06	0.17	1.05	0.21
$F_{s(5.5)}$	1.46	0.23	1.47	0.25
$F_{f(3.5)}$	1.24	0.19	1.13	0.23
$F_{f(4.5)}$	1.97	0.51	1.64	0.36
$F_{f(5.5)}$	2.89	0.68	2.24	0.39
$E_{(3.5)}$	1.40	0.38	1.28	0.28
$E_{(4.5)}$	2.15	0.49	2.16	0.50
$E_{(5.5)}$	3.32	0.71	3.76	1.18

Extension: The Friedman test showed a significant difference in movement time between control interfaces in each ID ($p < 0.001$, $nDOF = 3$, $\chi^2_{ID=3.5} = 16.2$, $\chi^2_{ID=4.5} = 24.36$, $\chi^2_{ID=5.5} = 24.36$). When ID = 3.5 (Fig. 8 and Table III), force on sternum was significantly faster than force on feet

and EMG. When ID = 4.5, force on sternum was significantly faster than force on feet and EMG. Additionally, the joystick was significantly faster than force on feet. Finally when ID = 5.5, force on sternum was significantly faster than force on feet and EMG. The Joystick was significantly faster than force on feet and EMG. Also, force on feet was significantly faster than EMG.

Since the differences between the control interfaces on the following metrics were not momentous, the average of all IDs was taken into account.

B. Throughput

Flexion: The Friedman test showed a significant difference in throughput ($p < 0.001$, $nDOF = 3$, $\chi^2 = 23.16$). In the Post-hoc analysis (Fig. 9 and Table IV), the force on sternum throughput was significantly greater than the force on feet and EMG. The joystick throughput was also significantly greater than force on feet and EMG. No significant difference was found between joystick-force on sternum and force on feet-EMG.

Extension: The Friedman test showed a significant difference in throughput ($p < 0.001$, $nDOF = 3$, $\chi^2 = 24.36$).

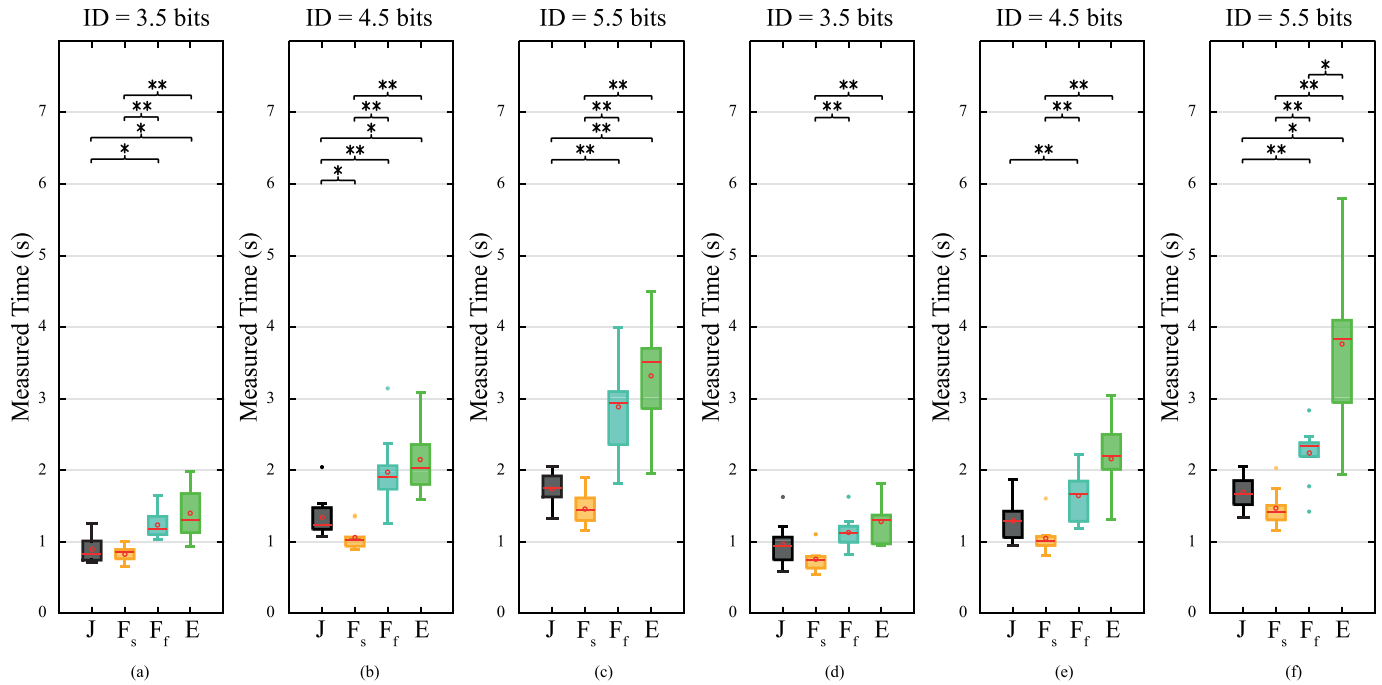


Fig. 8. Boxplots per ID statistics. (*) $p \leq 0.0083$ (**) $p \leq 0.0017$. Figures a – c correspond to flexion and d – f correspond to extension.

TABLE IV
TP AND STANDARD DEVIATION

	TP _{flexion}	SD _{flexion}	TP _{extension}	SD _{extension}
J	4.00	0.59	3.97	0.64
F _s	4.53	0.40	4.80	0.63
F _f	2.76	0.50	3.14	0.54
E	2.47	0.62	2.41	0.64

TABLE V
RT AND STANDARD DEVIATION

	RT _{flexion}	SD _{flexion}	RT _{extension}	SD _{extension}
J	0.55	0.09	0.48	0.07
F _s	0.56	0.06	0.50	0.05
F _f	0.61	0.11	0.61	0.15
E	0.65	0.06	0.60	0.07

In the Post-hoc (Fig. 9 and Table IV), the force on sternum throughput was significantly greater than force on feet and EMG. Joystick throughput was significantly greater than force on feet. No significant difference was found between joystick-force on sternum, joystick-force on feet or force on feet and EMG.

C. Reaction Time

Flexion: The Friedman test showed a significant difference in reaction time ($p < 0.05, nDOF = 3, \chi^2 = 9.84$). In the Post-hoc analysis (Fig. 9 and Table V), the only significant difference was found between force on sternum and force on feet with force on sternum being faster.

Extension: The Friedman test showed a significant difference in reaction time ($p < 0.001, nDOF = 3, \chi^2 = 21.36$). In the Post-hoc analysis (Fig. 9 and Table V), the joystick had a significantly faster reaction time than force on feet. Also, force on sternum had significantly faster reaction time than force on feet. No other significant difference was found.

D. Path Efficiency

Flexion: The Friedman test showed a significant difference in path efficiency ($p < 0.05, nDOF = 3, \chi^2 = 12.6$).

TABLE VI
PE AND STANDARD DEVIATION

	PE _{flexion}	SD _{flexion}	PE _{extension}	SD _{extension}
J	90.59	1.93	88.70	1.32
F _s	86.07	3.32	84.41	2.11
F _f	89.40	2.70	90.72	1.51
E	89.24	3.37	89.47	1.60

The only significance in this metric was between joystick and force on feet (Fig. 9 and Table VI).

Extension: The Friedman test showed a significant difference in path efficiency ($p < 0.001, nDOF = 3, \chi^2 = 22.68$). The joystick had a significantly greater path efficiency than force on sternum. Force on sternum had a significantly lower path efficiency than force on feet and EMG (Fig. 9 and Table VI).

E. Overshoot

In overshoot, the Friedman test showed that, for both flexion ($p = 0.067, nDOF = 3, \chi^2 = 7.17$) and extension ($p = 0.1236, nDOF = 3, \chi^2 = 5.72$), there was no significant difference between control interfaces (Fig. 9).

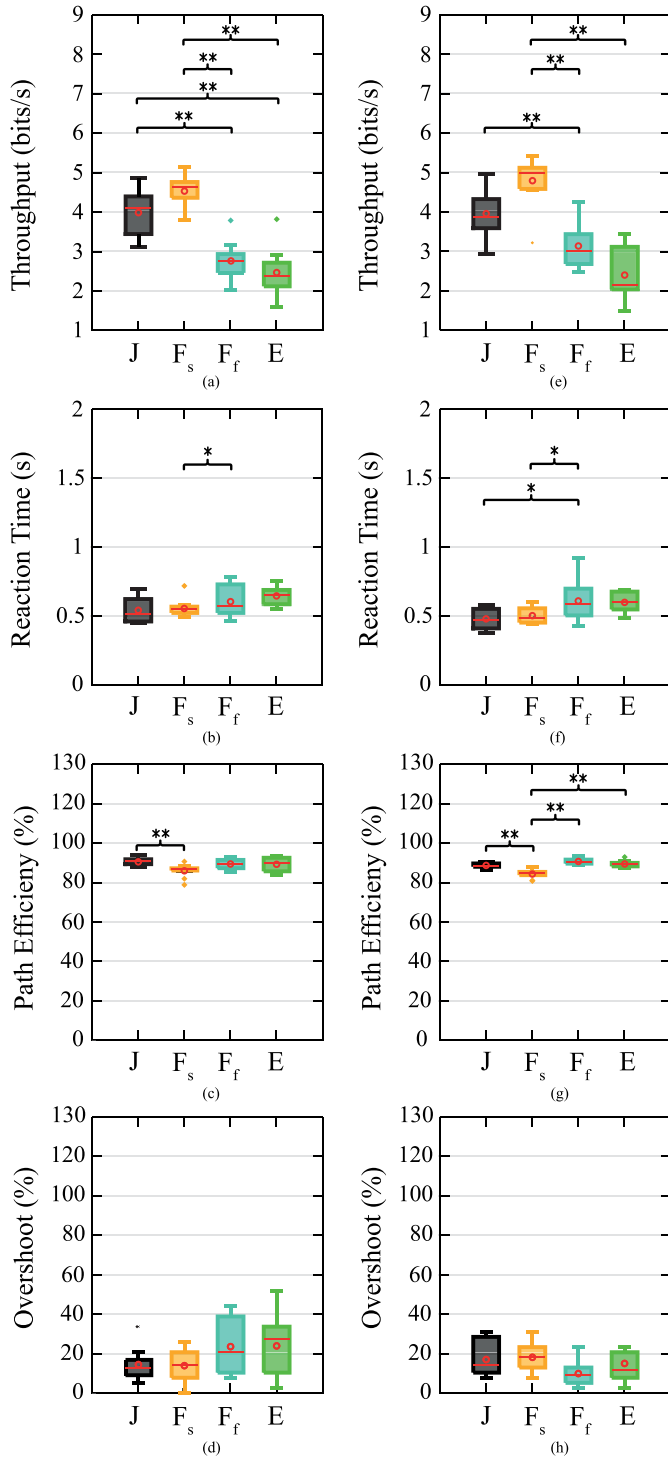


Fig. 9. Performance metrics statistics. Figures a–d correspond to flexion and e–g correspond to extension.

F. Questionnaire

Fig. 10 shows how the subjects replied to the questionnaire. The force on sternum was the fastest (6 out of 10), easiest to install (6 out of 10), comfortable (6 out of 10) but also the most exhausting (9 out of 10). On the other hand joystick was most accurate (6 out of 10) and easiest to control (7 out of 10). Regarding the overall performance, subjects favored the joystick (5 out of 10).

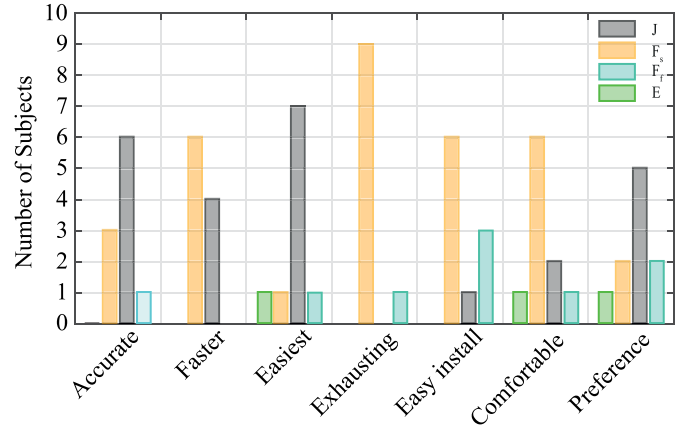


Fig. 10. Responses of the questionnaire.

V. DISCUSSION

This study investigated the performance of control interfaces on a novel active trunk support device. We designed and constructed a one DoF active trunk assistive device that can be controlled by four control interfaces. The four control interfaces were compared with the Fitts' law style experiments on performance using healthy subjects.

A. Performance Metrics and Acceptance

To evaluate the control interfaces, we considered time as an important factor because performing a task in a short amount of time indicates that the user has adapted to the control interfaces. Furthermore, accuracy performances such path efficiency and overshoot, compare the movement performance across the different control interfaces. Finally, the questionnaire gave a better insight into the control interfaces from the user perspective. The subjects did not have any previous experience with controlling assistive devices and were able to complete a series of trials with all the proposed control interfaces. These led us to the conclusion that the proposed control interfaces can be used for controlling a trunk support prototype.

B. Fitts' Law

Following Fitts' law, movement within the human can be described as an ID vs MT relationship. We used Fitts' law to investigate if the physical man-machine augmentation will disturb this relationship. R values higher than 0.5 would indicate a strong relation between ID and MT. Based on our r-values, it can be concluded that all four control interfaces follow this relationship between ID and MT. However, the regression coefficients of this study are lower than the ones found by others (<0.9) [20]. There are two possible reasons for this. First, the dwell time of the current experiment was used to identify the completion of the trial instead of tapping or pressing a button which changes the outcome of the experiment. Second, the control interfaces were used to flex and extend the human trunk which is not the same as performing fine movements with the hand.

C. Overall Performance

Based on the results of the MT, we can conclude that force on feet and EMG were slower than force on sternum and joystick for all IDs indicating that the subjects could achieve faster flexion and extension movements (attaining more than 70% in some cases). The MT results are in agreement with the performance of throughput where force on sternum and joystick have a larger transfer rate than force on feet and EMG. This result can be explained by the fact that the motion of the trunk can be controlled more intuitively by the joystick and the force on sternum. In contrast, force on feet and EMG from the leg muscle are considered to be less intuitive. The RT, PE differences were small (< 0.2 seconds, 4% respectively) which indicates only a marginal performance difference. Finally, there were no significant differences for the OS.

According to the subjects' responses to the questionnaire, the joystick was the easiest to control and the more accurate which contradicts with the results of the PE and OS metrics. The force on sternum was the fastest which is in line with the experimental results. On the other hand, the subjects found the force on sternum the most exhausting compared to the other 3 due to the fact that force signal contains both dynamic and static components of the upper body. Finally, the users' overall preference was the joystick which is not surprising because it is a very common control interface in, for example, video games.

D. Control Interfaces

The joystick was the first control interface to be used to allow the subjects to familiarize themselves with the system's dynamics. Hand- joysticks are commonly used as control interfaces for assistive devices (e.g., electrical wheelchairs) by individuals with muscular weakness and this was considered the easiest and most straightforward method to control an assistive device. Indeed, the joystick performed similarly to force on sternum although it was always the first control interface. The major drawback of using a joystick is that one hand's function is sacrificed whenever a person intends to move the trunk since it is a parallel system.

The force on sternum control interface was considered the most intuitive one. However, gravity compensation complicates matters as it acts on the upper body. Participants had difficulties in fully relaxing their muscles, which is important to achieve proper gravity compensation. Additionally, the interface was sensitive to respiration, which resulted in small oscillatory movements during the dwell time when the subjects were trying to keep their trunk at a certain angle.

The force on feet control interface was slower than the one using a sensor at the sternum, but gravity compensation was not necessary and the placement of the sensor not very critical. Although it is a parallel system requiring slight movements of the feet and it is slightly slower than the other control interfaces, it may be a solution for people who are seated in a wheelchair, because using the feet to control trunk movement is not a functional sacrifice.

EMG control performance was strongly dependent on where the sensor was placed on the muscles. It was easier to get a signal from the tibialis muscle to control the device. It was more difficult to find the right sensor location on the gastrocnemius so that the subject could control the device with ease. It should be mentioned though that at the beginning, the subjects needed some familiarization time to get used to the fact that they had to move their trunk by contracting their legs muscles. However, all the subjects were able to adapt to that procedure within minutes.

E. One-DoF vs Multi-DoF

The Peeters et-al study showed that the contribution of the trunk to achieve flexion tasks is divided equally between different segments (pelvis, lower lumbar, upper lumbar, lower thoracic, upper thoracic) [21]. As a consequence, an active trunk support should also be able to provide multi-DoF support, resembling natural reaching movements. Since no active trunk assistive devices exist, we decided to investigate the control capabilities of a relatively simple system before investigating a more complex assistive device. The DoF has to be increased for lateral bending as the current design restricts it.

An increase in DoF will introduce complexity not only to the mechanical design but also to the control. The mentioned control interfaces would have performed differently if the complexity of the control task had been increased.

F. Limitations

The first limitation is the setting of the admittance values which were tuned based on the performance on a pre-experiment with 2 subjects (not included in the results). The purpose of this pre-experiment was to find the optimal values of virtual mass and damping for each of the control interfaces. Even though the values of the virtual masses are close, the values of the virtual damping differ noticeably. This is due to the nature of the input signal in the admittance model. Control interfaces such as EMG need a bigger virtual damping value to attenuate the high input in the admittance model. Reducing the virtual damping value would result in a higher overshoot.

Second, the gear ratio backlash and the electronics of the motor resulted in mechanical play, giving 0 to 20 pixels in the GUI. Thus, not all the trials started at 0 or 1680 pixels (starting points for flexion end extension respectively), even though the subjects were asked to move the device to the mechanical end stops. We did not compensate for this non equality between trials since we considered it to be negligible.

VI. CONCLUSION

We investigated the performance of four different control interfaces on an experimental active trunk support device. The force on sternum and the joystick control interfaces were faster than the ones based on force underneath the feet and EMG. Regarding path efficiency, overshoot and reaction time, significant differences were found between the control

interfaces but the differences in absolute values are negligible. Force on sternum was experienced as the most fatiguing interface by the participants, and they preferred the joystick. From the above results, we can conclude that all four control interfaces can be potentially used to control an active trunk support with different advantages and disadvantages. Further research on the performance of the control interfaces will be done with people with DMD.

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